

Wireless and Non-contact ECG Measurement System – the “Aachen SmartChair”

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This publication describes a measurement system that obtains an electrocardiogram (ECG) by capacitively coupled electrodes. For demonstration purposes, this measurement system was integrated into an off-the-shelf office chair (so-called “Aachen SmartChair”). Whereas in usual clinical applications adhesive, conductively-coupled electrodes have to be attached to the skin, the described system is able to measure an ECG without direct skin contact through the cloth. A wireless communication module was integrated for transmitting the ECG data to a PC or to an ICU patient monitor. For system validation, a classical ECG with conductive electrodes and an oxygen saturation signal (SpO_2) were obtained simultaneously. Finally, system-specific problems of the presented device are discussed.

Keywords: electrocardiography, ECG, non-contacting, monitoring, capacitive electrodes, chair.

1 Introduction

Electrocardiography is one of the most important diagnostic methods to monitor proper heart function. Increasingly, it is not only used in a clinical environment but more and more applied to the “Personal Healthcare” scenario. In this application field, medical devices are intended to be used in the domestic environment and can communicate wirelessly [1]. Traditionally, for ECG measurement conductive electrodes have been applied which are directly attached to the skin. With the help of contact gel, they provide direct resistive contact to the patient. Unfortunately, these electrodes possess various disadvantages which are not optimal for long-term use in a “Personal Healthcare” scenario: as a result drying of the contact gel and surface degradation of the electrodes, the transfer resistance may change with time. Furthermore, metal allergies can cause skin irritations and may result in pressure necroses. Especially infant’s skin reacts sensitively to these kind of electrodes. Finally, as a single-use item, they are rather expensive. Capacitive (insulated) electrodes, which can obtain an ECG signal without conductive contact to the body, even through clothes, represent an alternative. This kind of electrodes were first described by Richardson [2]. Unlike conductive electrodes, the surface of these electrodes is electrically insulated and remains stable in long-term applications. Integrated in objects of daily life, they seem ideal for the “Personal Healthcare” field. Preliminary work concerning the integration of ECG measuring systems into objects of daily life was done by Ishijima [3]. In his work, conductive textile electrodes were used. These acted as underlay and pillow and could obtain an ECG during sleep. However, due to direct skin contact, the patient’s coupling was probably not exclusively capacitive. A group of Korean researchers around K. S. Park continued this work and integrated insulated electrodes into several objects, e.g. a toilet seat or a bath tub. Recently, two ECG applications with a chair have been presented [4, 5] that are somewhat similar to the work described below.

Thus, based on the measurement principles of insulated electrodes, we present a cable-free, battery-operated ECG measurement system, integrated into an off-the-shelf office chair.

2 Materials and Methods

Fig. 1 gives an overview of the developed measurement system. The front-end part of the measurement system consists of components for analog processing, like the insulated electrodes, an instrumental amplifier and analog filters plus in-between amplifiers. A/D conversion is followed by the digital part, with a radio transmitter as a data source and a receiver as a data drain. For demonstration and validation purposes, the receiver may be connected either with a standard ICU patient monitor via a D/A converter and magnitude adjustment, or with a PC via serial interface.

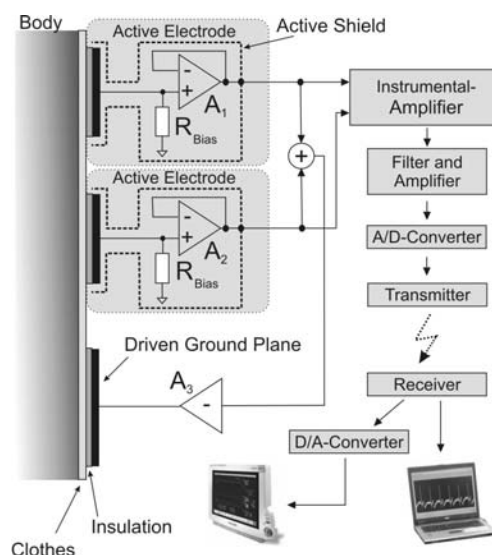


Fig. 1: Block diagram of the measurement system

2.1 Analog Signal Processing

The front-end features two active electrodes with an effective surface area of $A = 4 \text{ cm} \times 8 \text{ cm}$. In both cases, the electrode’s surface forms a coupling capacitance ($C_{1,2}$) between the subject’s body and the input of the unity gain amplifiers A_1 and A_2 , respectively (Fig. 1). The integrated unity gain amplifiers assure the required high input impedance of the

measurement system. The electrode's surface is covered by a very thin isolation layer (here: approx. 20 μm clear-transparent insulating lacquer). Hence, the coupling capacitance depends mainly on the thickness d and the dielectric constant ϵ_r of the cloth located between the electrode and the subject's skin. Assumed values like $d = 0.3 \text{ mm}$ and $\epsilon_r = 1$ result in a capacitance of

$$C_{1,2} = \epsilon_0 \epsilon_r \frac{A}{d} = 92 \text{ pF} \quad (1)$$

To suppress the interference due to the changing electromagnetic fields in the environment, the electrodes have to be actively shielded [6]. Any static charges on the coupling capacitances and/or on the subject's clothing must be discharged over the resistance R_{Bias} . Note that $C_{1,2}$ and R_{Bias} form a high-pass filter. A compromise between the discharging time constant and the attenuation of important ECG spectral fractions needs to be found by adjusting the resistor R_{Bias} properly. In this application, R_{Bias} was selected to $100 \text{ G}\Omega$. Thus, the cut-off frequency of the electrodes can be calculated to

$$f_C = \frac{1}{C_{1,2} \cdot R_{\text{Bias}} \cdot 2\pi} = 17.3 \text{ mHz} \quad (2)$$

In practice, any 50 Hz voltages from the power supply line that are capacitively coupled into the patient's body may cause a common mode voltage (V_{CM}) between the subject's body and the insulated circuit common ground V_{CM} of approx. 1 V, see Fig. 2. In this figure, C_S represents the isolation capacitance between the circuit common ground and the potential earth, and C_B as well as C_P the stray coupling with the 220-V-powerline and potential earth, respectively. Due to the finite common mode rejection ratio (CMRR) of the instrumental amplifier, these common mode voltages on the body cause interferences in the output signal V_{out} and have to be suppressed. For this reason, a so-called "driven-right-leg" circuit with an additional reference electrode is used in most conventional (conductive) ECG measurements [7]. By analogy, Kim et al. [4] introduced a so-called "driven-ground-plane" circuit for non-contacting ECG measurements which was also implemented in the application presented here. The sum of the electrodes' output signals is fed back via inverting amplifier A_3 to a conductive plane (C_3), as shown in Fig. 2. This plane is also insulated and capacitively coupled to the subject. Resistors R_4 and R_F adjust the amplifier's gain to

$$G_{A3} = \frac{2R_F}{R_A} \quad (3)$$

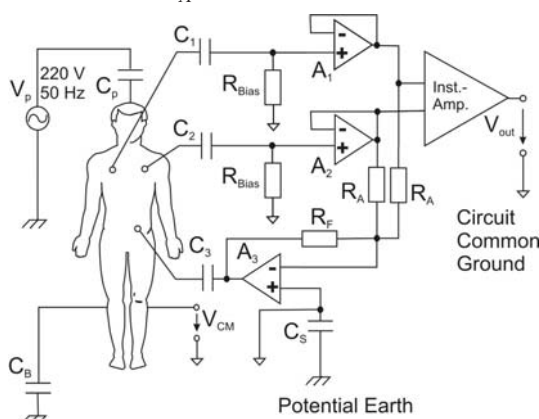


Fig. 2: The “driven-ground-plane” circuit

Besides the limited CMRR of the instrumental amplifier, another reason for needing to reduce V_{CM} is the possible transformation to a differential signal at the instrumental amplifier's input V_D , see Fig. 3. In the case of capacitive ECG measurements, this may occur due to different coupling capacitances $C_1 \neq C_2$, e.g. as a result of inhomogeneities in clothing.

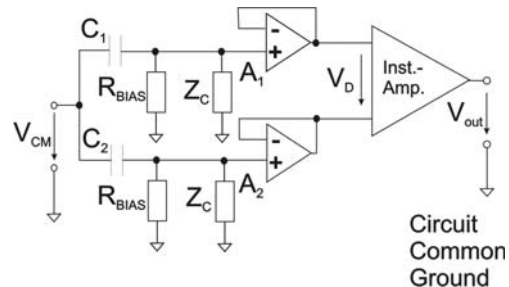


Fig. 3: Equivalent circuit for common mode voltages at the input of the measurement system

With $Z_1 = 1/j\omega C_1$ and $Z_2 = 1/j\omega C_2$ and the input impedance Z_C of A_1 and A_2 (here: OPA124, Burr Brown Corp., Dallas, USA), Eq. (4) shows the relation between V_D and V_{CM} :

$$V_D = V_{CM} \left(\frac{Z_2}{Z_2 + Z_B} - \frac{Z_1}{Z_1 + Z_B} \right) \quad (4)$$

with

$$Z_C = 10^{14} \Omega \parallel 3 \text{ pF} \quad (5)$$

and

$$Z_B = \frac{Z_C \cdot R_{\text{Bias}}}{Z_C + R_{\text{Bias}}} . \quad (6)$$

Furthermore, Eq. (4) illustrates that to minimize interferences at the instrumental amplifier’s input, either both coupling capacitances must be identical or V_{CM} must be as low as possible. A “driven-ground-plane” circuit supports the second option. The resulting common mode voltage suppression was found to be

$$\left| \frac{V_{CM}}{V_P(220\text{ V})} \right|_{f=50\text{ Hz}} = -118\text{ dB} \quad (7)$$

using a gain of $G_{A3} = 1000$ (compare with [4]).

Finally, the measurement chain for analog processing of the ECG signal consists of the following elements:

- Instrumentation amplifier (INA 114, Texas Instruments Inc., Dallas, U.S.A.)
- 4th order Butterworth high-pass filter with a cut-off frequency $f_{C,HP} = 0.5$ Hz
- An in-between amplifier
- 6th order Butterworth low-pass filter with cut-off frequency $f_{C,LP} = 200$ Hz
- 50 Hz Notch-filter with cut-off frequencies $f_L = 40$ Hz and $f_H = 60$ Hz

In the version presented here, the overall gain in forward direction (i.e. from the capacitive electrodes to the A/D converter, see Fig. 1) was set to 950.

2.2 Integration of the measurement system

The capacitive ECG measurement system was integrated into an off-the-shelf office chair. Fig. 4 shows the electrodes on the backrest and the position of the insulated reference electrode hidden under the cover (copper meshes as “driven-ground”, area: 34 cm×23 cm).

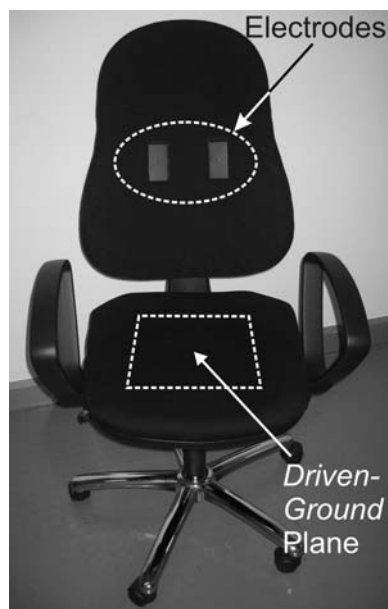


Fig. 4: The “Aachen SmartChair”, front view

The chair’s backrest is vertically adjustable, in order to adapt the position of the electrodes to the subject’s torso.

2.3 Wireless data transmission

The analog signal processing block is followed by a 12 bit A/D converter and a ZigBeeTM radio transmission module (ZEBRA module from senTec Elektronik Corp., Ilmenau, Germany). Low-power consumption and a data transmission rate sufficient for the specific application (unidirectional transmission of a one channel ECG) makes the ZEBRA module particularly attractive. A second ZEBRA module was used as a receiver. At the receiver’s output, two interfaces were implemented. Either an ICU patient monitor can be connected via D/A-conversion and magnitude adjustment, allowing standard ECG clamp leads to be applied, or a simple wired serial interface to a PC can be realized. In this second case, digital processing and the ECG data display may be done by a LabVIEW® application. Due to the wireless ECG data communication and the use of an accumulator for the power supply, the “Aachen SmartChair” possesses a mobility which is limited only by the transmission range.

3 Results

Fig. 5 shows an ECG signal obtained with our measurement system. All given measurements were recorded from the same subject (male, 26 yr., healthy) during normal breathing conditions and without further deliberate body movements. For further validation purposes, a classical, conductive Einthoven ECG and an oxygen saturation signal (SpO₂) were recorded in parallel to the capacitive ECG signal. These sig-

nals were displayed on an ICU monitor MP70 in combination with an “IntelliVue”-module (both produced by Philips Medical Inc., Boeblingen, Germany). As an example, a screenshot (inverted for better visibility) visually demonstrates the good correlation of the three vital signals (Fig. 6).

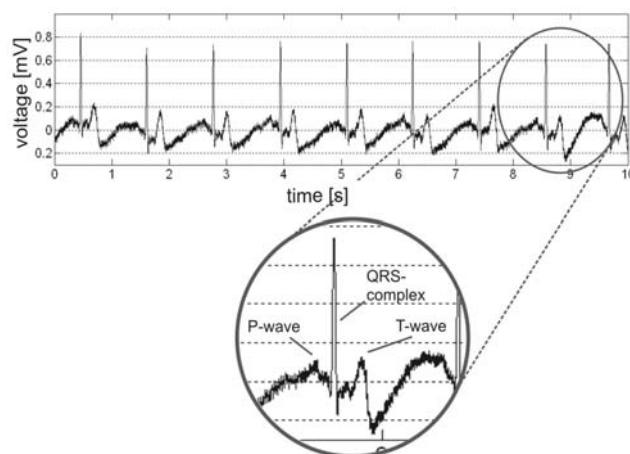


Fig. 5: Capacitively measured ECG while the subject is wearing a cotton wool shirt of 0.3 mm in thickness

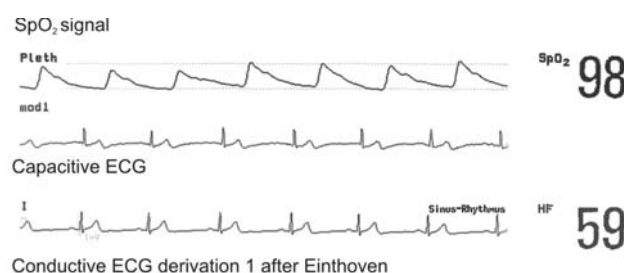


Fig. 6: Comparison between a synchronously measured SpO₂ signal (above), a capacitive ECG signal (center) and classical (conductive) ECG derivation after Einthoven (below). The ECG measurement was performed under normal breathing conditions, wearing a cotton wool shirt 0.3 mm in thickness.

In addition, Fig. 7 shows three capacitive ECG signals, obtained with our system, with different cotton wool shirt thickness.

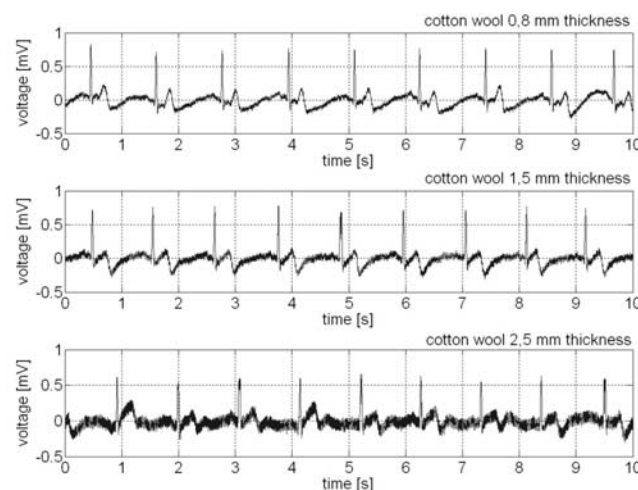


Fig. 7: Comparison of the ECG signals with different a cotton wool shirt thicknesses

4 Discussion

With our measurement system, it is possible to monitor an ECG without resistive skin contact. The QRS complex and the T wave were clearly identified, and often the p wave was also seen (see Fig. 5). However, typically a negative subsidence right after the T-wave, atypical in comparison to the Einthoven ECG, was observed by the authors, and this is in coincidence with the findings of Kim [5]. A possible reason for this could be the nontypical position of the insulated electrodes, compared to the classical ECG position. Also, this negative wave could be caused by a body movement, due to the mechanical activity of the heart. In the application presented here, the ECG electrodes are not fixed to the body. Thus, movement artifacts cannot be prevented in principle. A conductive ECG measurement may be less sensitive to movement artifacts, due to the use of electrodes glued to the body. To reduce movement artifacts, Kim et al. suggest raising the high-pass cut-off frequency from 0.5 to 8 Hz [5]. This method smoothes the base line, but may remove diagnostically important fractions of the ECG signal (like the P or T-wave).

Fig. 6 shows that the ECG signal quality of the presented measurement device is comparable to a conductive ECG measurement under certain clothing conditions: a subject's shirt thickness of approx. 0.3 mm or lower. Due to possible static charging of the clothes, a cotton wool material is preferred. After Searle [6], a general increase of the coupling impedance between body and measurement systems leads to an increased difference and, referring to Eq. (4), to increasing interferences in the ECG signal. As a result, Fig. 7 shows increasing 50-Hz hum. A reduced corner frequency of the low-pass filter to 35 Hz, also applied by Kim [5], could decrease this 50-Hz-noise. In any case, even with clothes thickness of 2.5 mm, at least the QRS-complex was clearly identified.

5 Conclusion

With the capacitively coupled ECG measurement system presented here, an ECG can be obtained without direct skin contact and, thus, without causing skin irritations. Compared to a conventional, conductive measurement system, it is more sensitive to moving artifacts. Furthermore, the quality of the capacitive ECG is strongly dependent on the subject's clothing, i.e. an adequate distance between the surface of the electrodes and subject's body is necessary for a high-quality ECG measurement. Taking these disadvantages into consideration, our system seems useful for heart rate detection in long-term applications. However, further research is needed

before the diagnostic potential of capacitive ECG measurement can be finally evaluated.

Acknowledgments

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